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FOR THE COMMANDER:

for CORNELIUS R. FAY III
Lieutenant Colonel, MS
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AD-B186 443

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(1)

CONTRACT NO: DAMD17-92-C-2050

TITLE: REAL-TIME, LIGHT WEIGHT, X-RAY IMAGER

PRINCIPAL INVESTIGATOR: Mustafa E. Kutlubay

CONTRACTING ORGANIZATION: Sensor Plus, Inc.
4250 Ridge Lea Road
Amherst, New York 14226

REPORT DATE: June 6, 1994

TYPE OF REPORT: Phase II - Midterm Report

PREPARED FOR: U.S. Army Medical Research, Development, Acquisition and Logistics Command (Provisional), Fort Detrick, Frederick, Maryland 21702-5012

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REPORT DOCUMENTATION PAGE	Form Approved OMB No 0704-0188

1. AGENCY USE ONLY (Leave blank)

2. REPORT DATE
June 6, 1994

3. REPORT TYPE AND DATES COVERED
Phase II - Midterm Report

4. TITLE AND SUBTITLE
Real-Time, Light Weight, X-Ray Imager

5. FUNDING NUMBERS
Contract No.
DAMD17-92-C-2050

6. AUTHOR(S)
Mustafa E. Kutlubay

7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)
Sensor Plus, Inc.
4250 Ridge Lea Road
Amherst, New York 14226

8. PERFORMING ORGANIZATION REPORT NUMBER
W104-MID

9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)
U.S. Army Medical Research, Development,
Acquisition and Logistics Command (Provisional),
Fort Detrick
Frederick, Maryland 21702-5012

10. SPONSORING/MONITORING AGENCY REPORT NUMBER

11. SUPPLEMENTARY NOTES
Phase II research for the period May 1, 1993 - April 30, 1994.

12a. DISTRIBUTION/AVAILABILITY STATEMENT
Distribution authorized to DOD Components only/
Specific Authority, June 6, 1994. Other requests
shall be referred to the Commander, USAMRDAL ,
Command (Provisional), Fort Detrick, Frederick,
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12b. DISTRIBUTION CODE

13. ABSTRACT (Maximum 200 words)

A digital X-ray imaging device is under development. The imager will have the capability to replace currently used x-ray film cassettes. The X-ray detection is achieved by means of fluorescent screen. An image formed on a fluorescent screen during exposure is captured by an array of charge-coupled-devices. CCD image sensors are read out by a high-speed, parallel channel design electronics. Overlaps between the image segments, as a result of array configuration, are removed, and the final image is displayed on a medium to high resolution display. For the purpose of portability, an LCD display and a compact size microcomputer unit are used in the system. However the system can be adapted to any other computer system or display unit with ease. The imaging system also includes number of image processing algorithms available to the operator for further refinement of the images.

14. SUBJECT TERMS
Radiography, Medical Imaging, Digital X-Ray Imaging,
X-Ray Detectors, Image Processing, Phase II

15. NUMBER OF PAGES

16. PRICE CODE

17. SECURITY CLASSIFICATION OF REPORT	18. SECURITY CLASSIFICATION OF THIS PAGE	19. SECURITY CLASSIFICATION OF ABSTRACT	20. LIMITATION OF ABSTRACT
Unclassified	Unclassified	Unclassified	Limited

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"Real-Time, Light Weight, X-Ray Imager"
Sensor Plus Inc.
Principle Investigator: Evren Kutlubay
Midterm Report

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1. Introduction

In the past years, there have been numerous techniques introduced for the representation of internal structure of human body, such as ultrasonography, magnetic resonance imaging, positron emission tomography, etc. However, radiography is still the most widely used method of medical imaging. In radiography, a beam of X-rays is passed through an object (human body) and received by a detector. X-rays are attenuated exponentially while passing through a medium. This can be expressed by the formula $I=I_0e^{-\mu x}$ where I_0 is the intensity of impinging beam, μ is the attenuation coefficient, and x is the thickness of a body. Since various components of a body have different attenuation coefficients, the detector receives different amount of radiation at each point corresponding to different attenuation. As a result, two dimensional information on the internal components of a human body is gathered.

Radiographic film is the most common conventional detector in which scintillation screen, together with photographic film, is used. The scintillation screen is used to convert x-ray photons to visible light photons. The film is exposed during the radiation by the illumination of the fluorescent screen. Then the film is chemically processed to obtain the resulting image.

Because of advancement in technology in the past decades, many other detection systems have been developed to be used as an X-ray receptor in medical imaging. For example, in phosphor computed tomography, images are captured in storage phosphor screens. The phenomenon, called laser-stimulated luminescence, is used to read out images stored. In this method phosphor plates are scanned by a laser beam, and following this photostimulation, the resulting light emission is detected by a photomultiplier. In fluoroscopy, X-ray image intensifier tubes are used. An image intensifier tube converts an X-ray into a visible light which is intensified while it is demagnified, and displayed on a phosphor screen for direct viewing. These intensifiers, recently, are coupled to television cameras by using lens or fiber optics. Phosphor screens can also be coupled directly to a charge-coupled-device (CCD) by fiber optic tapers. A fiber optic taper is a bundle of fiber optic cables bonded together. One end of a fiber optic bundle is squeezed so that the taper performs as a image reducer. At the front end of the taper phosphor is deposited for the reception of X-ray radiation and conversion to visible light emission. The other end, in most cases, is placed directly placed onto a CCD image sensor for image acquisition.

Each method used for X-ray detection has its own limitations. The film based systems require long processing time for the development of film and large area for storage. That is, images are not available immediately and digitization is necessary for further image manipulations and processing. Currently, sophisticated computer systems with extensive image processing software are a significant aid to medical diagnosis during the treatment of patients. For the systems using storage phosphor screens, images are also not immediately available and the cost of the scanners used in those systems is quite high. Image intensifying tubes are not suitable for portable systems. There is still a need for an additional system to capture and digitize the image acquired on a phosphor screen. Fiber optic tapers are, at present, limited to a size which is much smaller than the standard x-ray film sizes.

In this study there are several goals intended to overcome the limitations of the detection systems mentioned above. The goals are:

- To acquire a digitized image for further image manipulation and processing,
- To make the images available immediately to the operator during the treatment,
- To be compatible with conventional radiographic film size,
- To be portable for use in field hospitals,
- To have a dynamic range and spatial resolution comparable to conventional systems,
- To be cost effective,
- To develop a system which can easily be adapted as needed (change of detector area or change of computer).

An X-ray imaging system incorporating the above specifications is under development. It consist of a fluorescent screen for the conversion of X-rays to visible light, an array of charge-coupled-devices for acquisition of a image formed on a scintillation screen, and an image processing computer to reconstruct and display the images captured.

The fluorescent screen is chosen for optimal matching between its emission spectrum and spectral response of the photodetectors. It has the dimensions of a current radiographic film (8 inches by 10 inches).

Very high internal image resolution is achieved with a moderate quality optical system components and moderate resolution CCDs by combining subsection images from array of CCDs. The result is a 1300x1600 pixel image for 8x10 inch scintillation screen, which corresponds to a resolution of about 6 lines/mm. The spatial resolution of the imager can be increased by expanding the number of photodetector-lens pairs or by using higher resolution photodetectors, up to the spatial resolution of the fluorescent screen. By a unique parallel imaging channel design, a high resolution image is captured in a short time (seconds). The acquisition time can be decreased further by adding more parallel readout channels.

The software has been written for the reconstruction of the images captured from the detection unit. Each CCD provides one subsection of the image on the fluorescent screen. The computer aligns the subimages to reconstruct the complete X-ray image. With this method, image edges are <u>not</u> lost, as it is true for segmented fiber optic coupling.

The device can be used any branch of radiology, especially where high dynamic/spatial resolution and on-line verification is needed.

2. Description of the Imager (Current Status)

The first prototype is constructed and tested with an optical light/image source. It consist of the following sections:

- A fluorescent screen to convert X-rays to visible light. An optical source was used for the initial tests instead.

- Multiple image-sensor/lens pairs each capturing one segment of the image on the fluorescent/optical screen. The image sensor used is a 542 x 582 pixel charge-coupled-device (CCD).

- Data acquisition system to condition CCD's analog signal output and to obtain digital pixel information.

- Memory unit and communication interface to store the data gathered and to transfer the data to a computer system for further processing captured image.

- Computer system with LCD display.

- Touch screen for system/operator interaction.

- Alignment grid for the calibration of the detector.

- Reconstruction software for the alignment of the individual segments captured through the CCD/lens modules.

- Image processing software for further refinement of the images by an operator.

In the first half of the project period, all the components outlined above were purchased and/or fabricated. The system was tested with an optical source. A complete block diagram of the system with details is given in figure 2.

2.1 Fluorescent Screen

A commercially available Kodak Lanex intensifying screen is chosen for the purpose of X-ray to visible-light conversion. The selection of the screen was made on the basis of its spectral response, resolution and energy conversion efficiency. The spectral response of the Lanex screen begins at 415 nm and extends through 630 nm, with a peak response of energy located at 545 nm, which matches with the spectral response of the imaging sensor (400 nm and 600 nm). The specified resolution of the screen is 100 microns. Light output conversion efficiency for a 50 keV X-ray photon is approximately 1,000 light photons (within a factor of 2). System tests using this screen will be made in the next few weeks.

2.2 Image Detection Module (CCD/Lens Pair)

A Sharp LZ2324J CCD photodetector array is used as an image sensor. It has a 1/3 inch format solid-state imaging device with a 542 (H) by 582 (V) pixel array. Each pixel has a dimension of 9.6 μm horizontal and 6.3 μm vertical. The aspect ratio of 3:2 is corrected by software. The displayed image is not effected by the aspect ratio mismatch between the image acquired and the image displayed. Details of the driving electronics for the CCDs (Timer circuit) is given in Appendix A.

The lens design was made taking into account the light emission level of the fluorescent screen and overall size of the imaging device. At this point, the f-number, which defines the light collection efficiency of the lens system, can be adjusted from 1.2 and 16. The smaller the f-number the more the light is collected. The back focal length of the lens system is 8 mm and the image-to-object distance is 4 (four) inches. Further improvements on the lens system will decrease the image to object distance. Current the image distortion is about 1%. Distortion is defined here as the inability of the lens to image rectilinear objects accurately. Types of distortion include Pincushion and Barrel. Even though the distortion can be corrected by software, the goal is to minimize it by a lens system so that little or no additional processing time is needed.

2.3 Data Acquisition Unit

The data acquisition electronics starts with a CCD driving circuitry (Appendix A). The CCD is clocked to shift the charge accumulated in the photodetectors. Analog pixel information is shifted out serially as a voltage level. The method, called "correlated double sampling", is used to obtain absolute pixel charge information from the CCD output signal. The method eliminates the dark noise created by electron hole recombination in the photodetectors and removes the feed-through signals from the analog output. The resulting signal is then amplified and it's level is adjusted to match the signal converter. An 8 bit analog-to-digital (A/D) converter is used for the conversion of analog pixel information to digital data. The specific A/D was chosen because of its low error level (± 1/2 low significant bit) and high speed. Detailed diagram of the data acquisition circuit is given in Appendix B.

2.4 Memory Unit and Computer Interface

Each subimage, acquired through one CCD/lens pair, is stored in a memory (SRAM) as a image profile. Each image contains the data for the corresponding subsection of the complete picture on the fluorescent screen together with overlap from neighboring segments. In this application a 4 megabyte storage area is used for the complete unprocessed image. Appendix C contains the circuit diagram of the memory unit.

The detection unit is connected to the microcomputer unit. A microcomputer unit is used for reconstruction and processing of the acquired image. Currently an IBM PC-compatible 486DX microcomputer system is used for this purpose. The system is not limited to a specific computer. The interface card can be modified if any other (non-compatible with IBM PC) computer system is used. To achieve high speed, the data is transferred in parallel to the microcomputer unit's memory. One control register is provided in the interface card to allow individual access to any subimage in the detector memory. Appendix D contains the circuit diagrams of the units.

2.5 Liquid Crystal Display (LCD) and Touch Screen

One of the requirements for the system is portability. Therefore a small size and low power consumption are immediate concerns. A medium resolution monochrome LCD display was chosen because of its size. The microcomputer unit and the LCD display are mounted in a compact package. The resolution of the display is 640 by 480 pixels. There are 32 grey levels but dithering may be used to achieve a higher dynamic range.

As the user interface, a high resolution touch-sensitive screen is mounted on the LCD display. The operator interacts with the imager and microcomputer through the touch screen, and all the features of the software are accessible. The resolution of the touch screen is 1024 by 1024.

2.6 Reconstruction Software

After the image acquisition and transfer has been completed, the captured image is displayed on the user's LCD screen. Twelve full images are captured simultaneously by the hardware unit. After transferring these images into memory, the overlap between figures is removed and a single image reconstructed.

Currently a test screen is employed to rapidly determine regions of overlap, which can then be employed for image reconstruction. It is expected that such a test screen would be employed on a periodic basis to allow for unit recalibration. The effectiveness of such a test screen approach has been demonstrated previously, during the Phase I development. The current test screen approach has a significant speed advantage over many more complex registration approaches. If the larger number of image segments are employed, which will complicate the simple test screen method, then a Chamfer matching registration approach will be employed instead. Chamfer matching allows for the determination of the rotational and translational parameters where the image must be shifted to be in a one-to-one correspondence with a base image. Chamfer matching approaches rapidly extract binary edge information from the images to be matched and then determine the optimal match via an energy minimization scheme.

2.7 Image Processing Software

A full featured image processing software system (XRAYWIN) has been designed, taking into account the specific requirements of the X-ray imager. XRAYWIN provides a software interface to the X-ray imager unit's hardware as well as methods with which to improve a physician's visualization of a captured digital radiograph. XRAYWIN has been written for the Microsoft Windows operating system, affording the end user a high degree of user-friendliness and system flexibility.

The most important aspect of the design of XRAYWIN is its interaction with the X-ray imager hardware unit. The hardware unit was designed to operate semi-autonomously, without any closed loop control from the controlling computer. The hardware unit may receive and process a set of fixed commands which include instructions to acquire an image set and memory bus switching commands. When an image is to be captured, XRAYWIN sends an acquire signal to the hardware unit. After a sufficient time has elapsed to allow for image capture, the acquired data is transferred from the hardware units memory into the controlling computer's onboard storage area. XRAYWIN reorganizes the data as it is transferred in order to reconstruct the odd and even frames that have resulted from the interlaced CCD devices. This data shuffling process has been incorporated into the data transfer operation so as to eliminate the processing overhead that might otherwise be associated with the use of inexpensive, interlaced CCDs.

After image reconstruction the final, high resolution image is displayed for the clinician. The LCD unit currently in use cannot simultaneously display the entire image due to resolution limitations of such display devices. XRAYWIN allows the user to interactively scroll through the image, zooming in and out whenever necessary. XRAYWIN also provides the user with a variety of image analysis and enhancement tools with which enhance their visualization of the acquired digital radiograph. An example of the XRAYWIN user interface is presented in figure 3.

XRAYWIN provides the end user with several spatial filters which can be used to enhance image features of interest. Low pass filters have been implemented for image smoothing and noise reduction. A high pass filter function, which sharpens edge information, has also been implemented. A user-selectable filter option has been included for the more advanced user. This option allows the user to arbitrarily select filter coefficients for a 3x3 convolution filter.

3. Results and Discussions

The imaging device was completely tested by using an optical source and a test screen. The system was demonstrated to be able to acquire high resolution images. The device captures twelve (3 by 4) images on a back-illuminated screen, which simulates a fluorescent screen. Sub-images are acquired through CCD/lens combinations. Each CCD sensor stores one related sub-image in its photodetector array as a charge accumulation. The charge is transferred out by a high speed electronics circuit which is partially controlled by a microcomputer. Images are digitized and stored in a memory as a image profile. Then the information is accessed by a microcomputer unit for processing and display. Various image processing algorithms are available in the system for further evaluation of the images. The image capture time is about 400 milliseconds. The total time for the reconstruction of the complete image and display takes about 7 seconds (on a 486 PC).

Because there are overlaps between the subimages, the information at the edges of the sub-sections is not lost. The amount of the overlap on a test screen is used to calibrate and align the device. Since there will be differences between the devices manufactured due to tolerances of the PCB layouts, mechanical construction, CCD positions and lens configuration, the system should be calibrated or aligned by special test screen(s). Moreover the system needs to be calibrated periodically to correct for possible deformation of the packages due to use or temperature change. The next prototype of the device will have isolated CCD modules (CCD with lens) mounted on a sturdy frame so that at the time of manufacture, each CCD module can be adjusted mechanically (Figure 1). This procedure will eliminate the small rotational errors due to CCD die/PCB mounting, and positional errors between the CCD and lens mount. Additionally, the modular configuration was selected because of its suitability for mechanical reconfiguration. Each module will have its own signal conditioning and digitization units so that the modules can be added together to form a much larger imaging area. Appendix E contains the circuit diagram of the module. Each module will run in parallel to increase the acquisition speed so that X-ray exposure time can be reduced. Advancement of fluorescent screen technology is also expected to further reduce X-ray exposure. No modifications to the electronics circuits will be needed. The only modification needed to add modules will be to reconfigure the interconnections between the modules and/or units.

The captured image (Figure 4) shows all twelve (3 by 4) sub-images. There is some distortion present, which can be seen close to the edges of the images. The distortion with this lens system appears to be a barrel distortion. The distortion, defined as separation of the actual image point from the paraxially predicted location on the image plane. Here it is expressed by a percentage of the paraxial image height and is about 1%. The distortion can be corrected by a software operation, but this will increase the processing time. Therefore, we prefer an improvement on the lens design. Also study of the optimum correction of such a distortion by a software procedure is under way.

There is an aspect ratio mismatch between the image sensor array and the image display. The system corrects the CCD's 3:2 aspect ratio to display unit aspect ratio of 1:1. After the application of the correction algorithm, the resulting image can be seen in the figure 5.

Alignment of the sub-images begins by finding the cross-sections of the lines on the specific test screen. Since each image has the information from the neigbouring sub-images, each line cross-section is seen by four sub-sections. The algorithm matches those crossing points and the resulting image is the reconstructed complete image of the screen. However, because of the distortion mentioned earlier, some of the lines are not perfectly aligned and curvatures can be seen at the edges. We are evaluating whether a change in lens design or application of software procedure is the most efficient method of distortion correction. Figure 6 shows an aligned image. Different test screens will be viewed to find most efficient pattern for the system. In the final device, the calibration screen will be used such a way that it will be viewed during an X-ray exposure. It will not be necessary to remove the florescent screen for calibration. Also the procedure will provide more information about the dynamic range of the system and corresponding grey levels on the screen. As a result, parameters for the contrast enhancement (if necessary) are obtained.

The dynamic range of the system first prototype is low because of the noise on the modified and patched circuit boards. During the test of the first prototype, preliminary circuit boards were used. The board layouts are far from optimum for updated electronics design. The second prototype will have the desired dynamic range of 1000:1, using the redesigned boards. Certain minor modifications are expected to reduce the noise further.

The imaging system already has the spatial resolution comparable to currently used digital radiographic systems. The system has the resolution of 1300 horizontal and 1600 vertical pixels, corresponding to about 6 lines/mm. The resolution can be increased further by using higher resolution CCDs or by increasing the number of CCDs modules used in the same imaging area. However there will be a maximum resolution which is limited by the fluorescent screen. The resolution of the fluorescent screen, in general, is limited by the grain size of the X-ray phosphor, and the thickness of the phosphor deposited. System tests with an X-ray source and a fluorescent screen are about to start. As mentioned earlier, Lanex screens, which have the resolution of 10 lines/mm, were chosen for these tests. The system resolution is comparable with the resolution of the fluorescent screen. Investigations of improved screens will continue in the coming months.

The portability of the device is an important requirement. Because the microcomputer has a relatively high power consumption, it may have to be replaced. However, by altering some of the components on the system, such as hard drive, the power consumption can be reduced. The advancements in the area of portable computers will be followed to find best unit available.

Preliminary steps to find an industrial partner for commercialization of the X-ray image were made, particularly during the last three months. Sensor Plus is looking for a larger company with substantial financial resources both for the manufacturing of the Phase III portion of the SBIR and also for development and marketing of civilian products.

IV. Conclusion

The digital X-ray imager is of particular value in areas of medical radiographic imaging where on-line verification of the results are needed. Further, portability of the system will allow the operator to use the imager in field hospitals or mobile units.

The imaging device under development is able to capture high resolution images without loosing information present on the screen, that is, image information at the edges is not lost due to recombination of the subimages, as true for systems based on fiber optic tapers. The resulting radiographic image has the standard film size used in medical radiography. The dynamic range of the first prototype is mediocre but will be improved in the second prototype. It will have both dynamic range and resolution equal to the currently available comparable digital imaging systems. The segmented imaging technique, with its image reconstruction software has, the acquisition speed to rapidly display radiographic images after exposure. The speed can be increased by adding more parallel channels into the system. This unique technique allows the system to be modified easily for the applications requiring different imaging areas.

In the following months a second prototype will be constructed. The new prototype will provide low noise levels by using optimally designed printed circuit boards. The mechanical construction of the device will be changed to provide more rigid mounting. During the construction of the second prototype, the first prototype will be used to test the response of the system to an X-ray exposure. Lanex screens will be used for the initial tests. Phantoms will be available to be used under X-ray exposure.

Additionally, in the coming months, advancements in the area of intensifying screens and portable computer systems with a high resolution LCD displays will be followed and improvements incorporated into the new design.

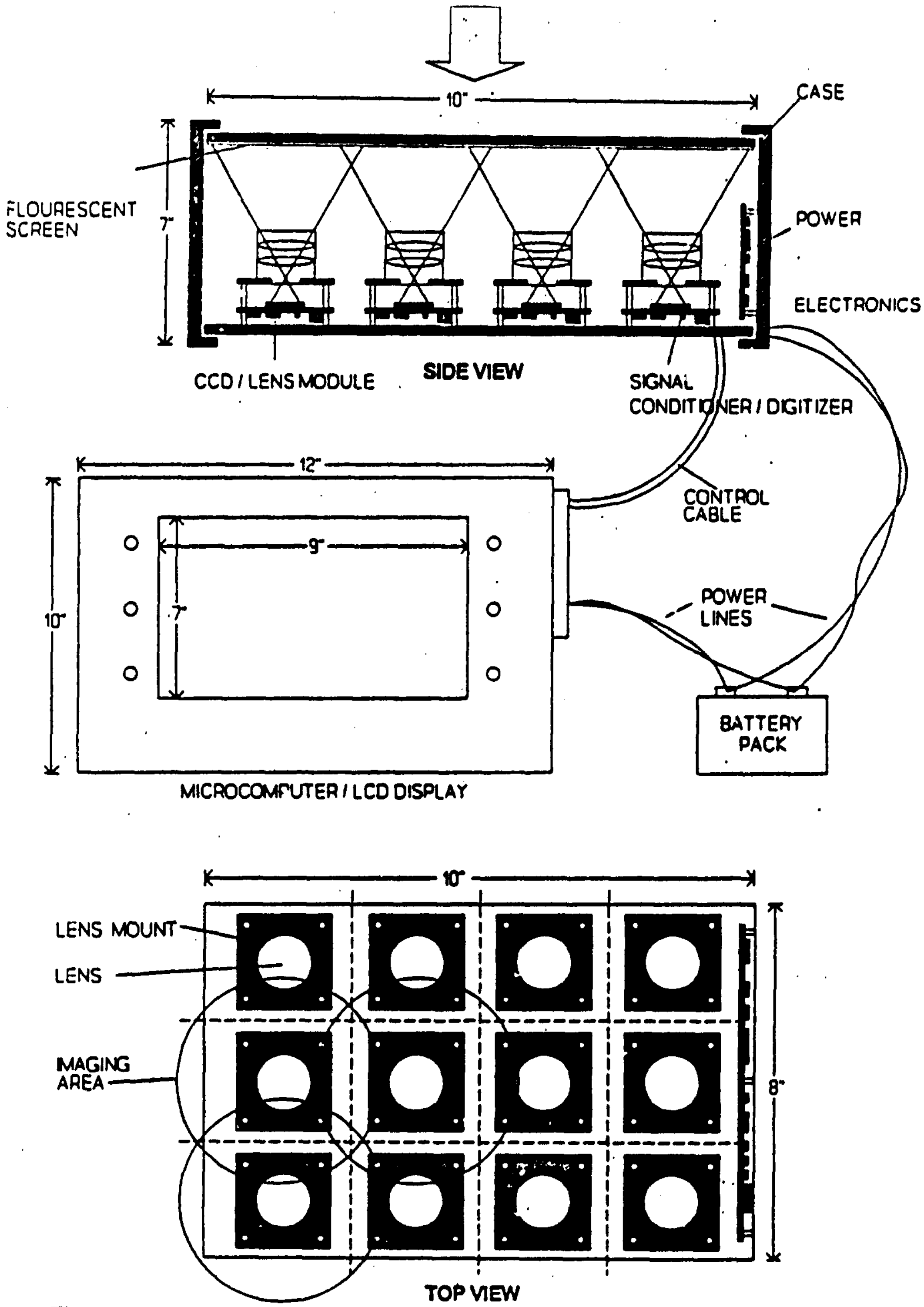

Figure 1. The X-ray Imaging system.

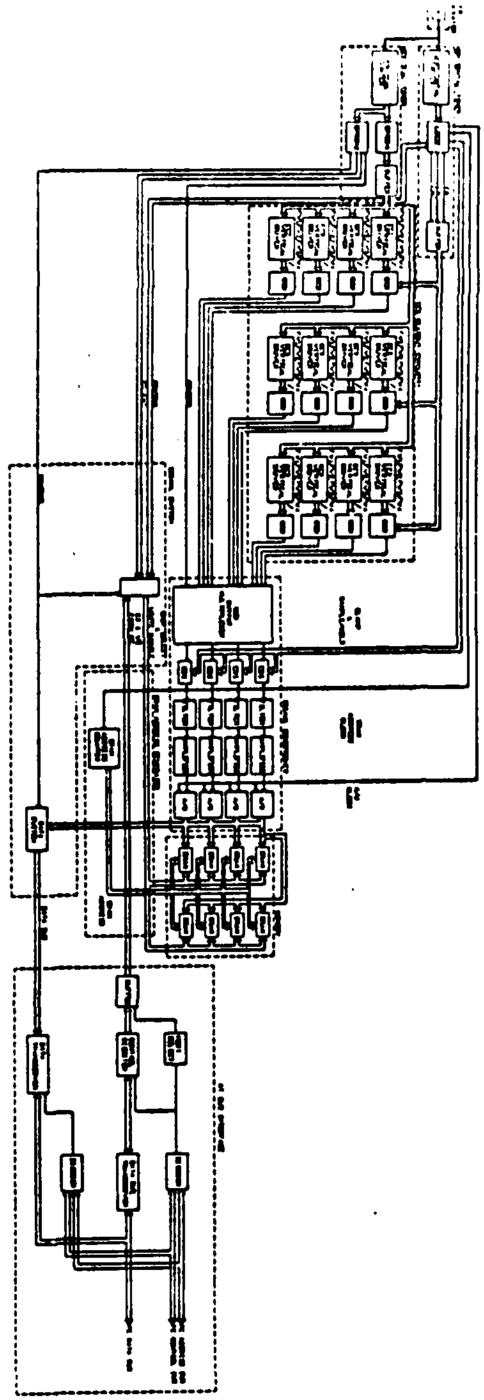

Figure 2. Block diagram of the electronics.

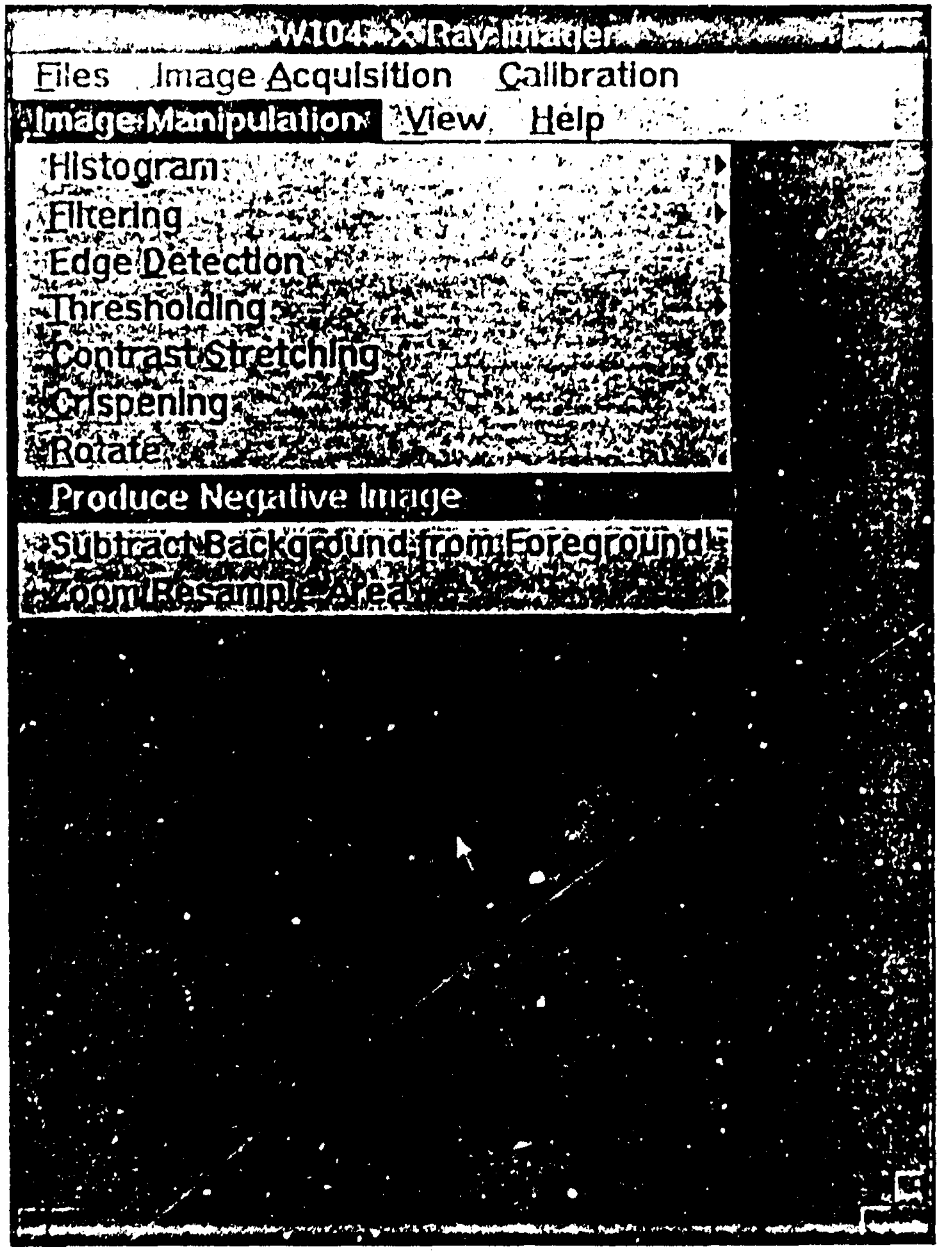

Figure 3. An example of XRAYWIN user interface.

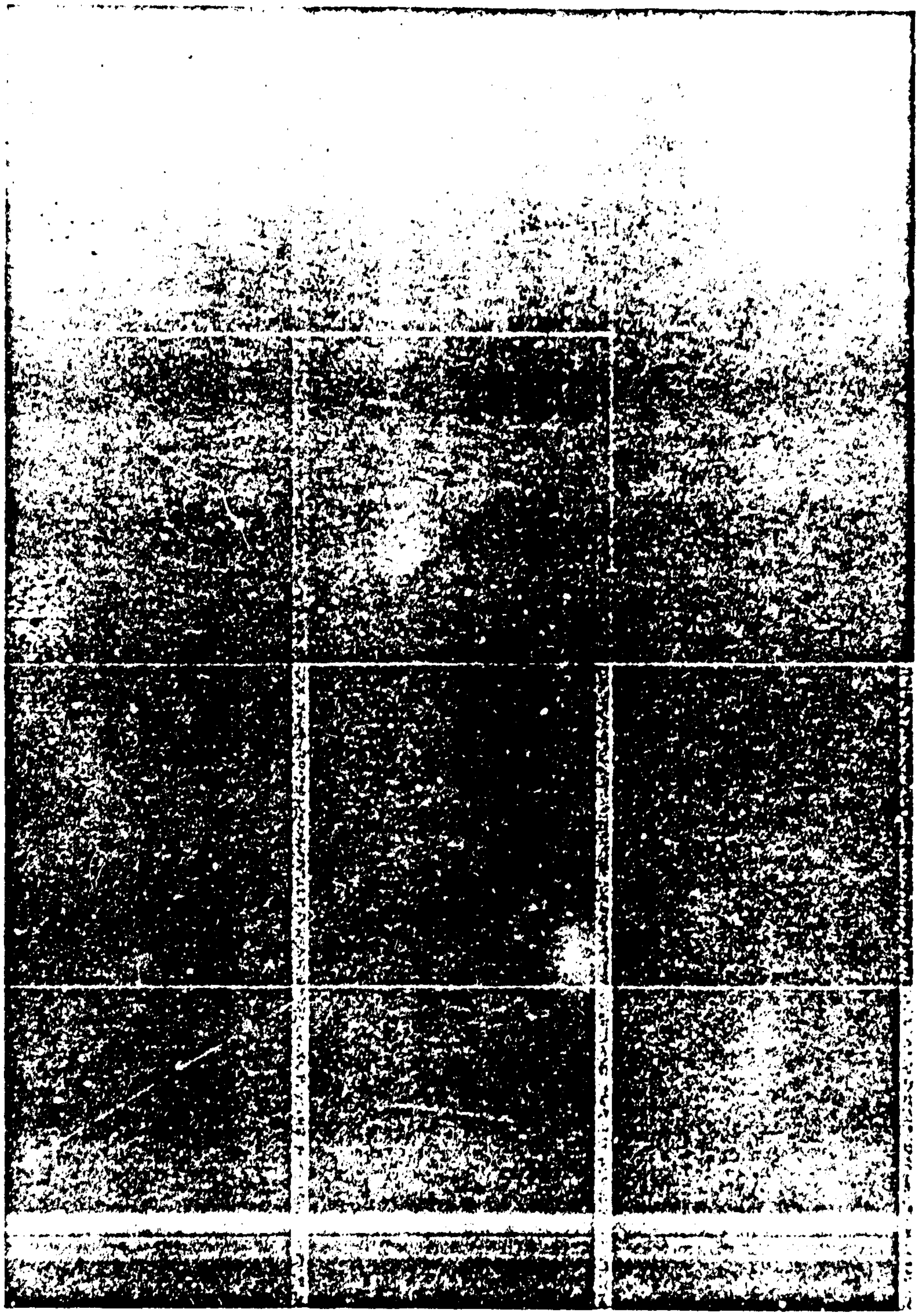

Figure 4. Acquired image before reconstruction.

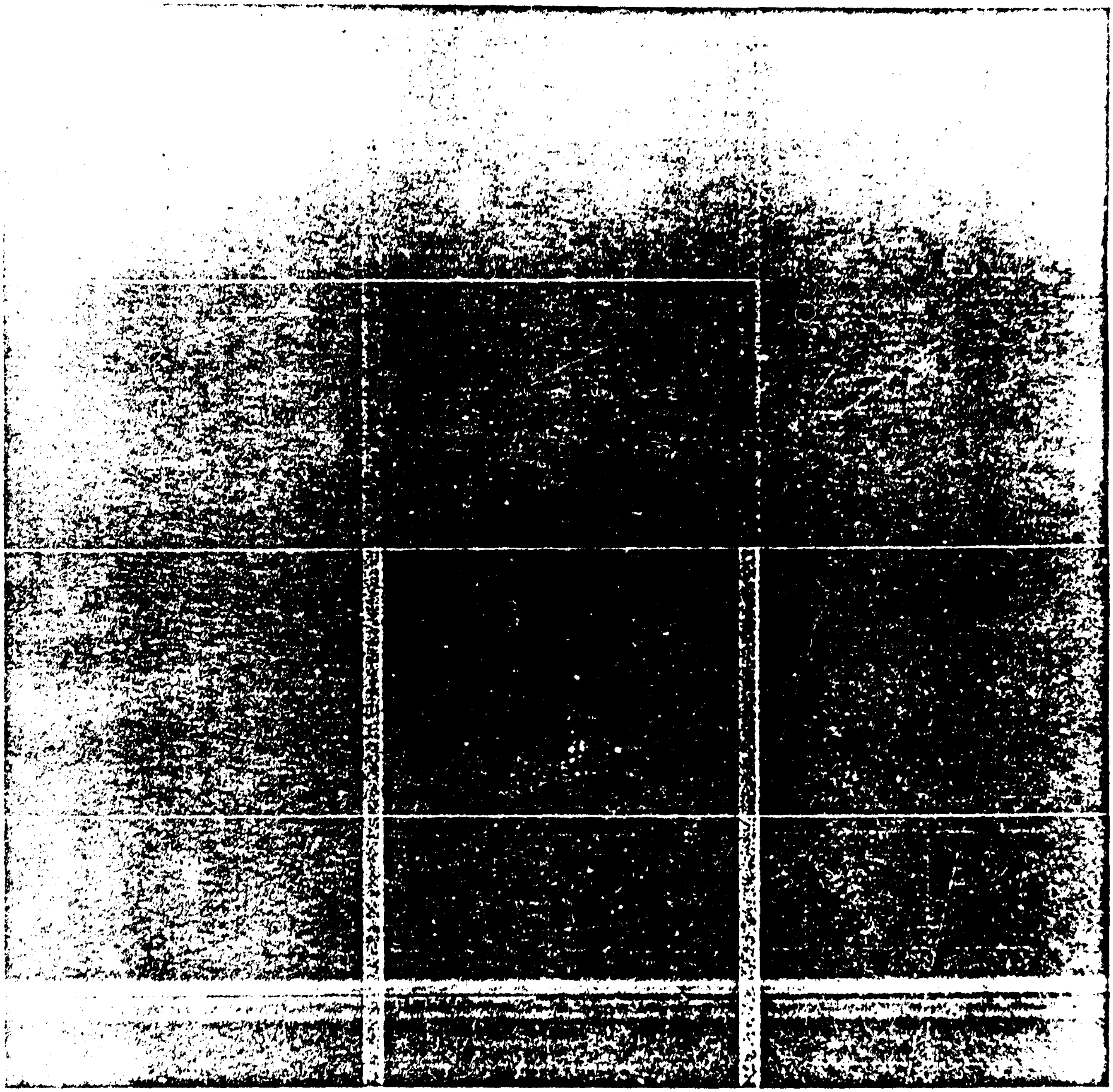

Figure 5. Image after aspect ratio correction.

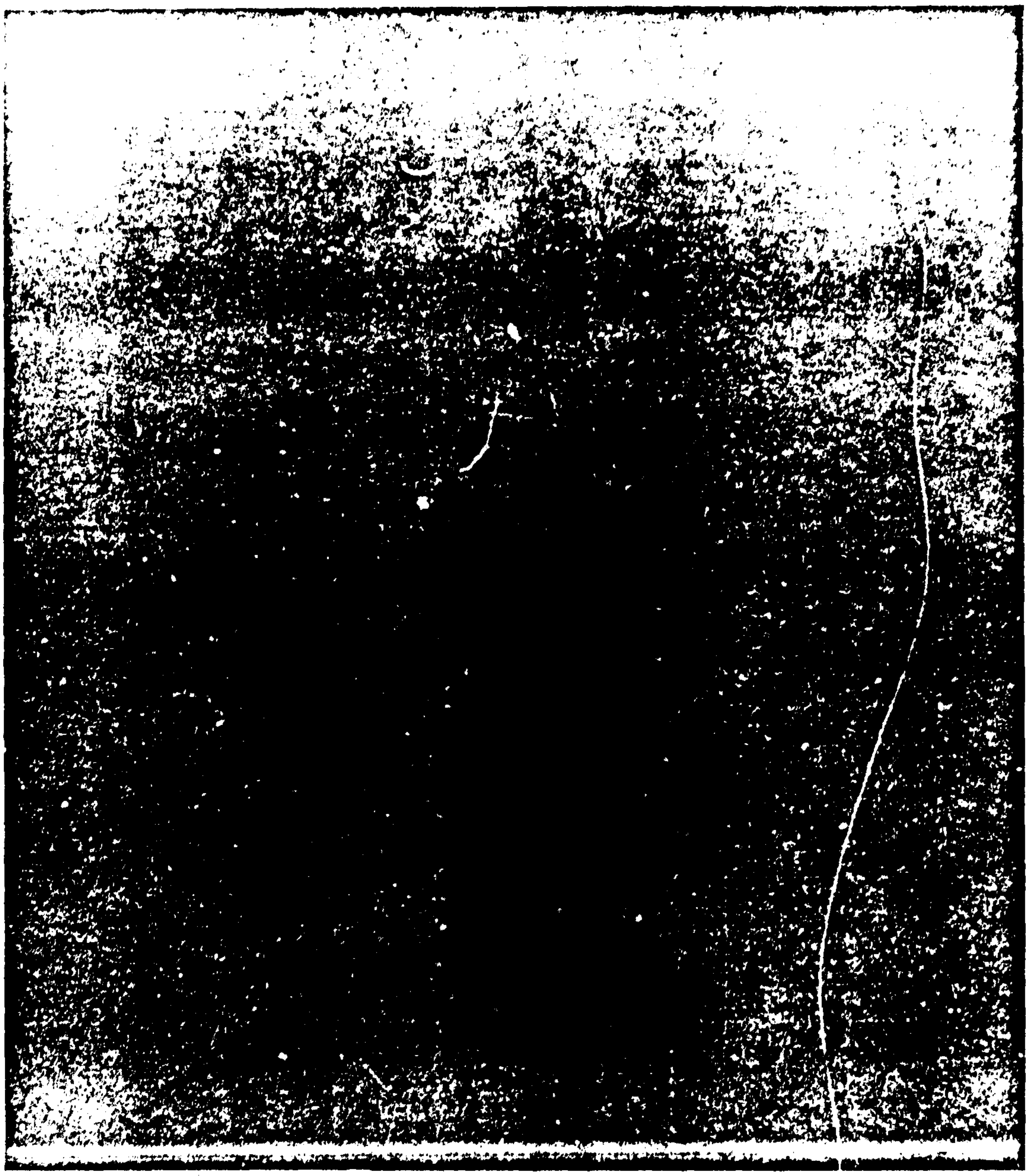

Figure 6. Reconstructed image.

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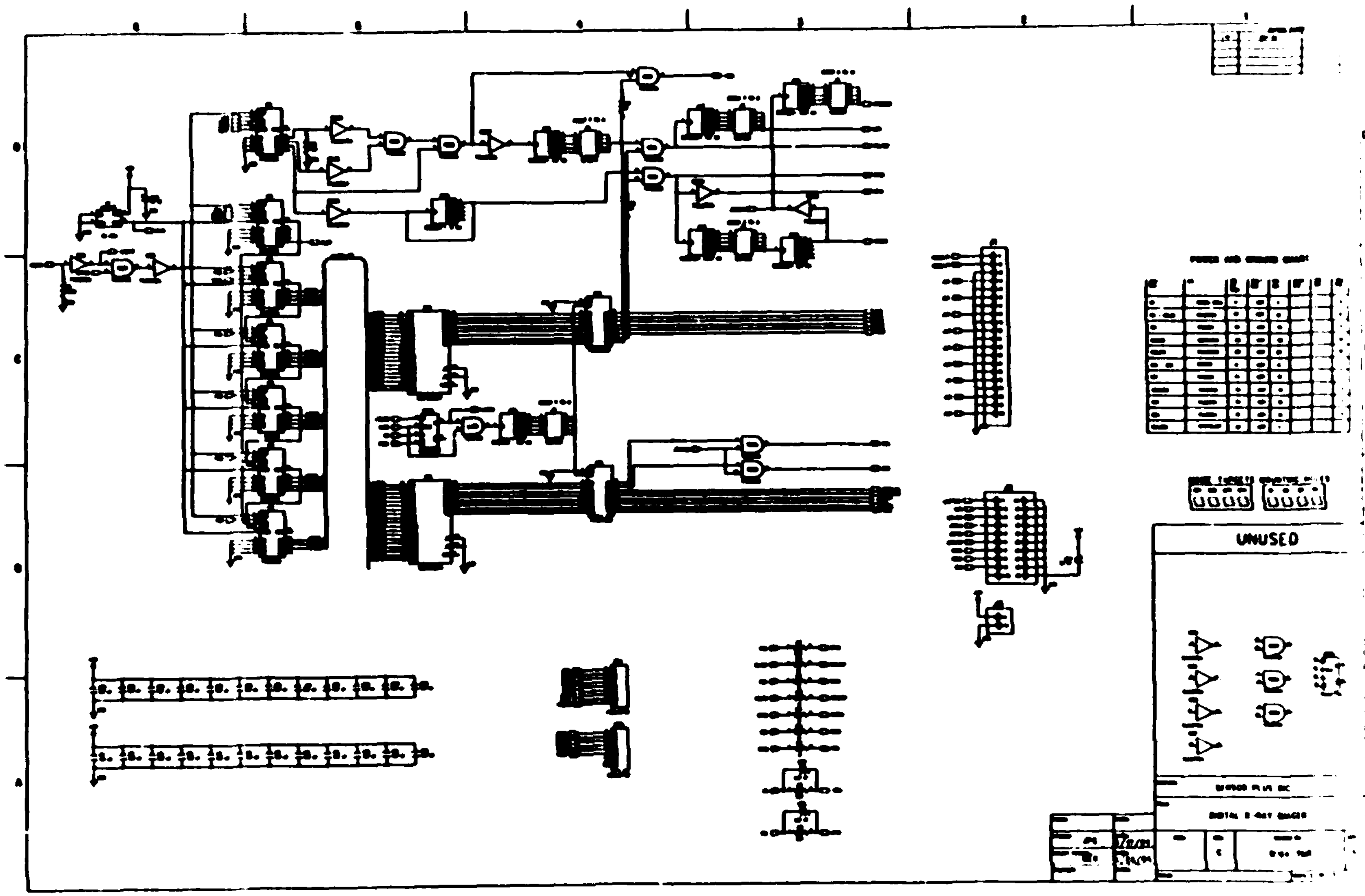
Appendices

(For the full copies of the diagrams contained in the next pages, contact the Principle Investigator of the project at Sensor Plus Inc.)

Appendix A.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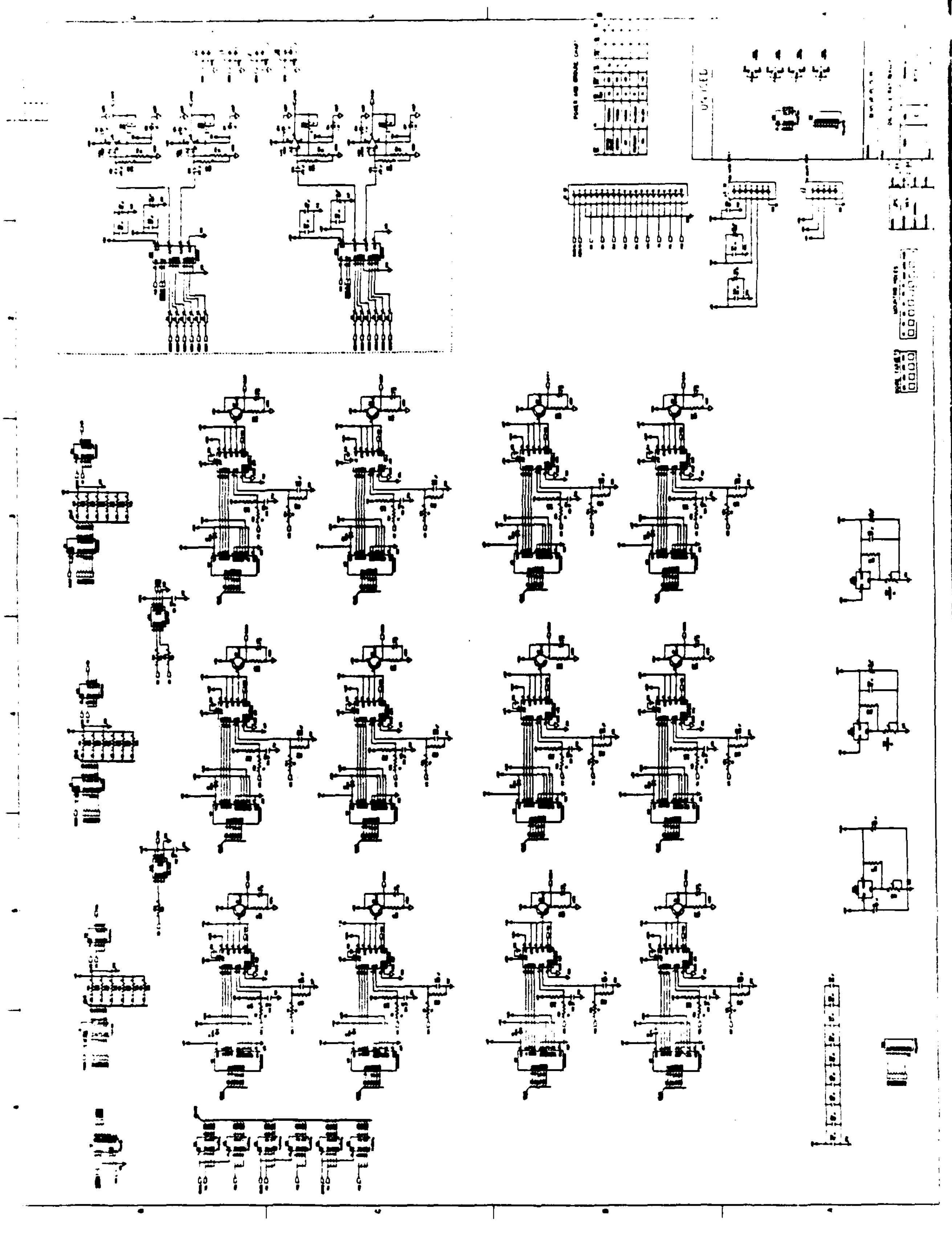
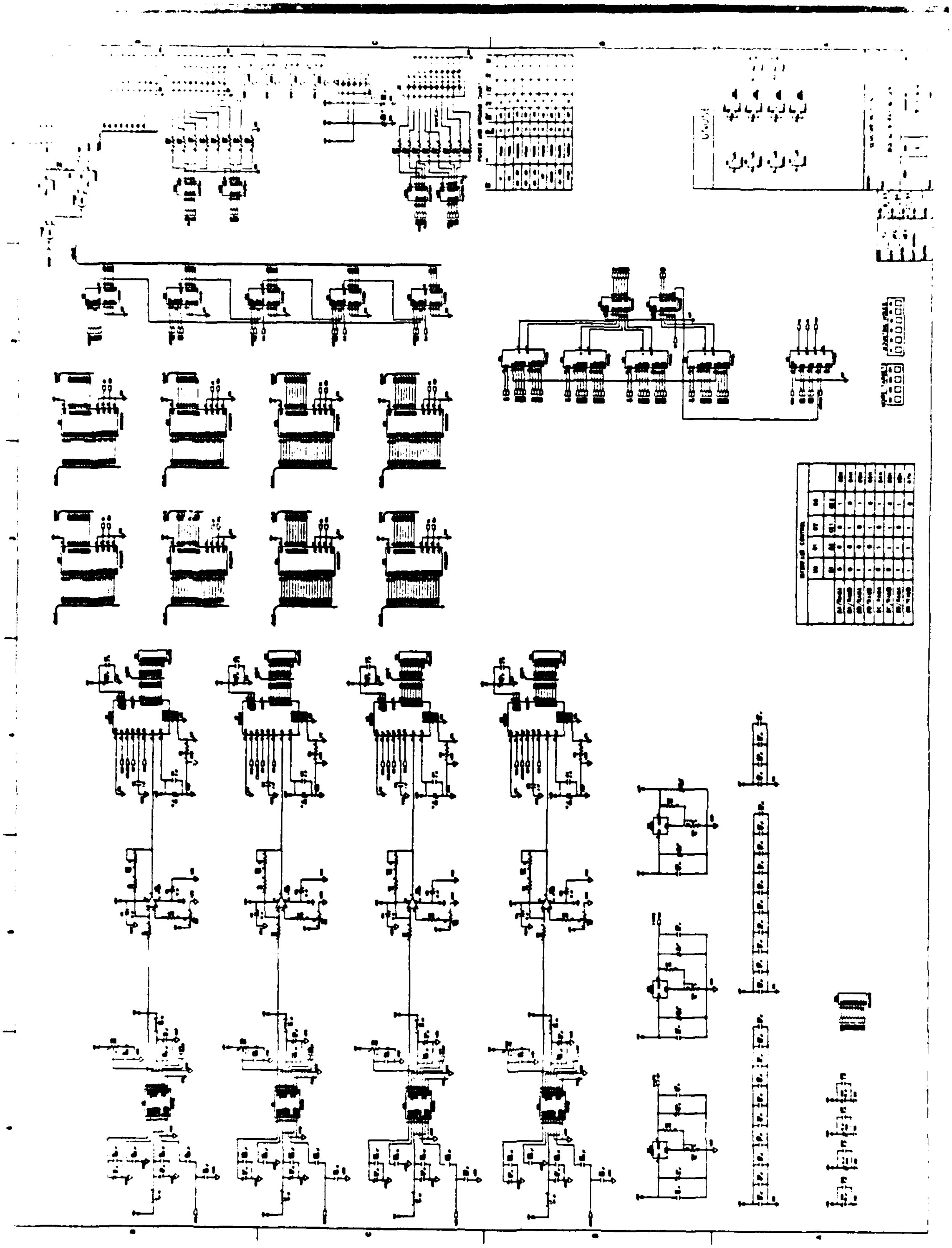
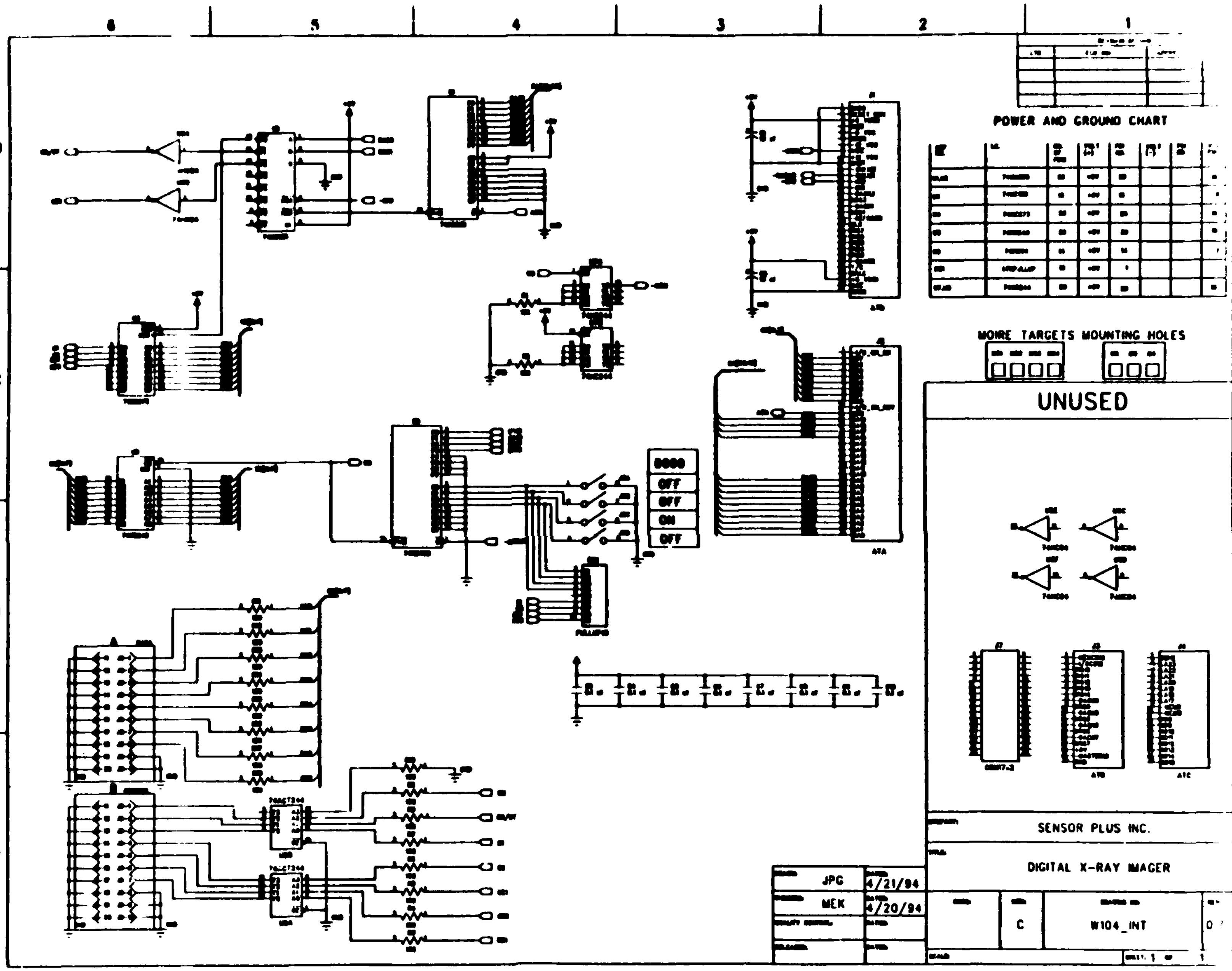
Appendix B.



Appendix C.



Appendix D.



POWER AND GROUND CHART
MOIRE TARGETS MOUNTING HOLES
UNUSED
OFF
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SENSOR PLUS INC.
DIGITAL X-RAY IMAGER
JPG
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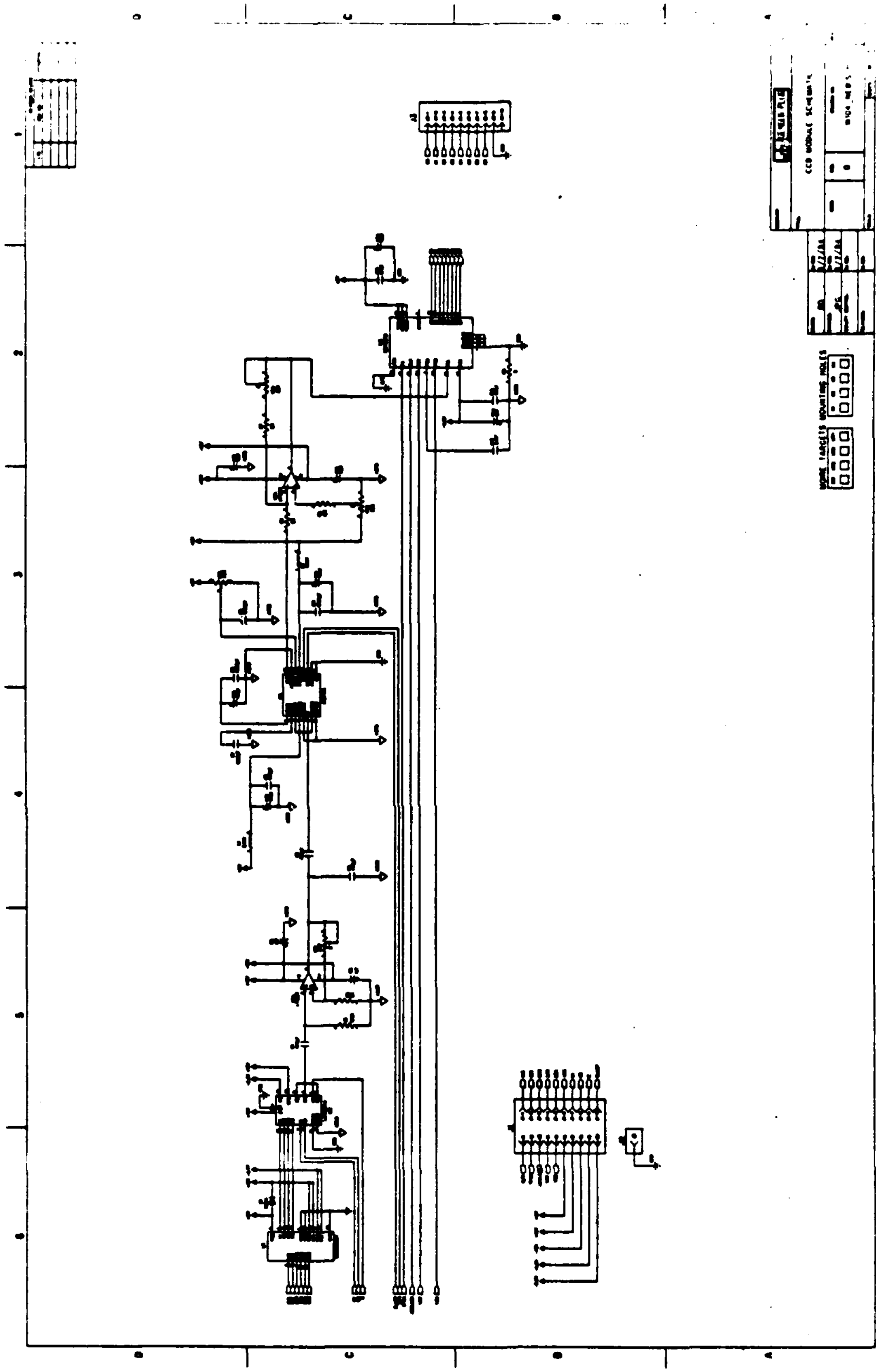
Appendix E.


CCD MODULE SCHEMATIC
MOUNTING HOLES